ISBN: 978-3-95450-215-8 ISSN: 2226-0366 doi:10.18429/JACoW-LINAC2022-MOPOGE10

A MEDICAL LINAC FOR AFFORDABLE PROTON THERAPY

S. M. Hunt, Joël Adélise, Elsa van Garderen, Wolf-Dieter Klotz, Rebecca Seviour[1]
Alceli Limited, Aberdeen, Scotland
Dora Correia, PSI, Villigen, Switzerland
[1]also at University of Huddersfield, Huddersfield, UK

Abstract

Proton Therapy (PT) was first proposed in the 1940s. Application of this knowledge was largely led over the next fifty years by accelerator laboratories, but now also by commercial companies. Availability of PT is increasing but is limited by three factors: facility size, prompt/induced radiation, and treatment cost. Compact cyclotrons/synchrocyclotrons for single-room facilities have reduced space requirements. Linacs can avoid high radiation levels. Yet treatment costs have remained stubbornly high, driven largely by maintenance and staffing costs over the typical 20-30 year facility lifetime. Current technology cannot simultaneously reduce these three factors. By using a long Linac, the Alceli approach sacrifices size limitations, to gain massive improvements in treatment cost and radiation levels. Quadrupling the length of a Linac results in a sixteen-fold reduction in RF power per cavity. Along with other innovations in our design, this leads to a modular warm Linac with distributed solid-state RF amplification, easy and cheap to manufacture and maintain, requiring no water cooling, and a treatment cost of 1/10th of current facilities, making PT much more affordable.

WHAT IS PROTON THERAPY?

Traditional radiotherapy uses X-rays, a form of high energy electromagnetic radiation, to kill cancer cells. The X-rays pass through the body, depositing energy as they do so, and this energy kills both cancer and normal cells on the way. The beam is quite large, but by rotating the beam around the patient, and always pointing at the tumour, the tumour receives the maximum dose, and other tissue receives a lower, although non negligible, dose. Proton Therapy (PT) is an advanced form of radiotherapy that can treat tumours with minimal damage to the surrounding tissue. It uses protons instead of X-rays to destroy the cancer cells. The advantage of using protons is that instead of destroying cells all the way as they pass through the body most of the energy is deposited, and therefore damage occurs, at a specific depth known as the Bragg Peak. The depth of the Bragg Peak is dependent on the kinetic energy of the proton, and therefore can be controlled.

HISTORY OF PROTON THERAPY

The idea of using this effect for treating cancer was already proposed in the 1940's by Wilson [1], and first experiments treating patients were made in the 1950's at the Lawrence Berkeley Laboratory in California using their cyclotron [2] However, the high cost of the accelerators needed to accelerate protons to the necessary energy meant that for many years Proton Beam Therapy was restricted to accelerator laboratories treating small numbers of patients. The first dedicated accelerator to treat patients was a synchrotron designed and built by Fermi lab National Laboratory in the USA and installed at the Loma Linda Hospital in California in the 1970s [3].

Commercial companies only became involved in Proton Beam Therapy in the 1990's. The first commercial accelerator for PT was ordered by the Massachusetts General Hospital in the USA [4]. They were already making use of the Harvard Cyclotron to deliver PT and wanted their own dedicated machine. The contract to supply the machine was won by IBA, a Belgian company that already produced small cyclotrons used for science and other medical purposes. This one order has led to PT now being IBA's main business, and they have become the world leader in producing PT systems. Soon after, the Paul Scherrer Institute (PSI) in Switzerland which already had a very successful PT program treating patients using protons from its large cyclotron used primarily for high energy physics, ordered a superconducting cyclotron from Accel AG [5], a German Accelerator company. Having a dedicated accelerator for PT meant patients could be treated all year around rather than relying on the availability of the physics machine. Accel gained enough knowledge of PT from this project to offer machines to other sites, and Accel was eventually sold to Varian and has become a major supplier of PT systems including two NHS facilities in UCLH London, and the Christie in Manchester. So, from these two commercial contracts the PT industry was born. It is interesting to note that the prevalence of cyclotrons (and the related synchrocyclotrons) as accelerators for PT came as a result of these two contracts from centres already using cyclotrons for other purposes.

LIMITATIONS TO THE WIDE AVAILABILITY OF PT

Three factors have held back the wide adoption of PT world-wide.

Cost

The cost of treatment using PT is partly due to the high capital cost of building the facilities (circa £100M for the Christie in Manchester, significantly more for UCLH in London). But as that capital cost can be amortised over a long lifetime of the facility, even more important is the reoccurring cost of operation and maintenance of the facility (circa £20M per year for the Christie and UCLH).

Content from this work may be used under the terms of the CC BY 4.0 licence (© 2021). Any distribution of this work must maintain attribution to the author(s), title of the work, publisher, and DOI

ISBN: 978-3-95450-215-8 ISSN: 2226-0366 doi:10.18429/JACoW-LINAC2022-MOPOGE10

Content from this work may be used under the terms of the CC BY 4.0 licence (© 2021). Any distribution of this work must maintain attribution to the author(s), title of the work, publisher, and DOI

Radiation Produced

The cyclotrons typically currently used for PT generate high prompt and induced radiation. This is due to the unavoidable high losses at extraction (often >15%) and the fact that these losses are at the maximum energy (typically 230 MeV to 250 MeV) as the cyclotron is a fixed energy machine. These losses cause the unavoidable radio-activation of the cyclotron over time. Also, the degrader needed to reduce the proton energy to the level required for treatment generates high radiation. These high levels of prompt and induced radiation require large amount of concrete shielding, and more importantly potentially very high costs of disposal at end of life.

Footprint

Compared to traditional radiotherapy using a compact electron Linac to produce X-rays, proton therapy facilities have at least an order of magnitude larger footprint. Typical facilities comprising a cyclotron, RF sources, amplifiers, transmission lines, power supplies, beam lines, gantries, and shielding, require thousands of square meters of floor space. Some 'compact' single room solutions, often using synchro-cyclotron technology go some way to reducing this space requirement - but still require a specialised multi-story building.

An Ideal Accelerator for Proton Therapy

Despite the radiation produced, cost, and size of existing facilities - they are very powerful tools that can have a very important role in treating specialised hard to treat cancers, for patients who have very limited options. Examples of this are the treatment of certain brain cancers which otherwise have a very low survival rate, and paediatric cancers as children are more susceptible to long term damage from radiation. Although currently expensive, PT can be very cost effective in these cases.

IS THIS THE RIGHT APPROACH?

The ideal accelerator for PT would have:

- **Low cost** to build, operate and maintain.
- **Low radiation** produced.
- **Small size**.

Unfortunately, this is not possible. Remember "The perfect is the enemy of the good". Attempts to make a 'perfect' accelerator for proton therapy will fail. Linacs can have very low losses and therefore low radiation. But attempts to make them shorter means much higher RF power is needed, vastly increasing cost and complexity. Also attempting very high operating frequency leads to problems and cost in production to achieve the very high tolerances of cavities and alignment necessary. Currently, approaches such as Dielectric Wall Acceleration, have not succeeded in this domain due to enormous power needed and very complex technology.

OUR SOLUTION

As we can't achieve all three design goals simultaneously, the question is which one do you drop? Our approach is not to worry about size as we consider the two most important features are a very large reduction in radiation and large reduction in cost per treatment. We have chosen a Linac as the best way to achieve very low losses and therefore radiation. But what length of Linac? A typical Linac to accelerate protons to 200 MeV would be about 50 m long, but if we were to halve that length we would need to double the acceleration (field) per unit length, which would mean with the same shunt impedance four times the RF power needed. But we choose not to halve the length, but to double it (1/4 RF power per cavity), then double it again (1/16th RF power per cavity). In this way although cavity costs increase linearly with length, RF costs reduce with the square of the length.

Our Cavities

Our patented design consists of thousands of very simple cavities with loop couplers, that are extremely cheap to mass produce. Each cavity needs only 600 W peak RF power to produce the necessary field of approximately 1 MV/m and has its own semiconductor RF requiring only a single chip power amplifier costing £100. Each cavity also needs individual phase control to enable operation at variable proton energy. This distributed RF scheme eliminates the need for Klystrons and waveguide distribution, lowering costs and increasing reliability. As we operate at 1% duty cycle, average power per cavity is just 6 W - eliminating the need for water cooling, which improves reliability, simplifies maintenance, and reduces cost. Our cavity design was achieved initially using the simulation tool Superfish for 2D design, using the Finite Element based Comsol and Finite Integration based CST for 3D design which included loop coupler design and time dependent simulation of EM fields, and transient heating.

Operating Frequency

We have chosen 800 Mhz as an operating frequency, with a cavity diameter just under 30 cm as a compromise between smaller diameter cavities at higher frequency and engineering problems and cost of RF amplification if frequency is higher.

Our Lattice

Most of our cavities are 4cm wide, with ten cavities making up the most common cavity module designed to be quickly replaced if necessary. Each of these modules has a 10cm gap between them for either quadrupoles, vacuum connection, or diagnostics. Our FODO lattice is stable between 6 MeV injection up to between 80 MeV and 200 MeV maximum energy, and the system uses permanent magnet quadrupoles rather than electromagnets, so reducing construction and operating costs significantly, whilst offering increased reliability.

Modelling

We developed our own modelling/simulation code based on scientific Python that we named **SIMULINAC**. This code comprises of two Python3 modules to simulate a proton beam in a Linac. **simu.py** calculates beam envelopes in

ISBN: 978-3-95450-215-8 ISSN: 2226-0366 doi:10.18429/JACoW-LINAC2022-MOPOGE10

linear approximation according to Courant/Snyder theory as shown in Fig. 1.

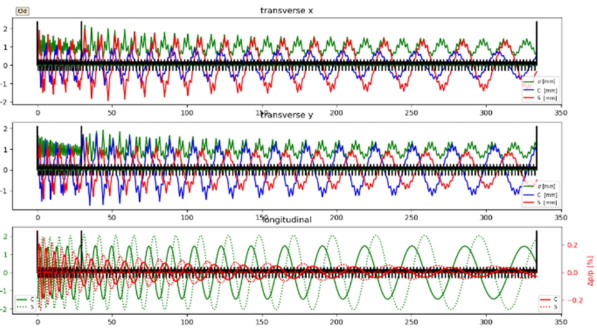

Figure 1: Lattice.

Tracker.py is a module that tracks a bunch of particles through the same lattice using linear approximations for the focussing elements but using different non-linear mappings for the traversal of accelerating RF-cavities. The emittance **ε** is a conserved quantity in linear theory and is a measure of the area of the ellipse in phase space. From the emittance of the beam and the twiss parameters at the entrance of the lattice the transverse beam envelopes **E(s)x.y** are calculated, see Fig. 2, along the lattice taking into account the *relativistic* increase of the impulse of the accelerated protons.

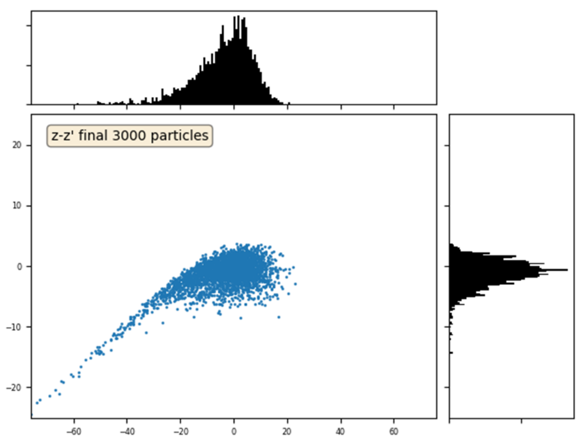

Figure 2: Final longitudinal emittance.

RF-cavities in the lattice are replaced by **drift-kick-drift** triplets, where the kick is a linear matrix or non-linear mapping approximating the acceleration in the cavity gap. The details of the linear kick matrix are described in the Trace3D documentation [6]. We also implemented non-linear cavity models[7]. We test the result of our code against other linac-codes freely available on the Internet[8, 9].The optimal length of our Linac is approximately 300 m, but we loop back at a fixed energy of 80 MeV halfway along so the dipoles can again be permanent magnets. The accelerator building length is approximately 160 m, but only 2 m wide, therefore overall, very little land is needed. We will produce and commission the Linac in 13 shipping containers in a factory, ready to ship and install quickly on site which requires only a simple concrete base and power connection.

Production

Our cavity and overall design prioritises low construction and operating costs over maximising performance. This makes it very suitable for producing an accelerator from 6 MeV to 200 MeV at a low cost, in quantities of more than ten a year, but more importantly at very low operating and maintenance cost. Along with the Ion source, RFQ, and treatment delivery, our systems are capable of treating up to 1000 patients a year (if 20 fractions per patient) at a cost which is up to an order of magnitude less than current PT technology.

CONCLUSION

We have designed and produced prototype cavities and RF amplifiers of a variable energy proton Linac able to significantly reduce the cost of proton therapy delivery, and so make it more available world-wide, without the problem of dealing with handling of radio-active components at end of life.

REFERENCES

[1] Wilson, R R, “Radiological use of fast protons”, *Radiology*, vol. 47, p.487-91, 1946.

[2] Lawrence, J H, “Pituitary irradiation with high-energy proton beams: a preliminary report”, *Cancer Res.,* vol. 18, pp.121-34, 1958

[3] G. Coutrakon *et al.,* “A performance study of the Loma Linda proton medical accelerator”, *Med. Phys.*, vol. 21, pp.1691-1701, 1994.

[4] Y. Jongen *et al.*, “Progress Report on the Construction of the Northeast Proton Therapy Center (NPTC) Equipment”, in *Proc. PAC'97*, Vancouver, Canada, May 1997, paper 7P103, pp. 3816-3818.

[5] Schillo, Michael *et al.,* “Compact superconducting 250 MeV proton cyclotron for the PSI PROSCAN proton therapy project”, *AIP Conf. Proc.*, vol. 600, pp.37-39, 2001.

[6] Trace3D, `https://laacg.lanl.gov/laacg/services/traceman.pdf`

[7] Shishlo & Holmes, `https://info.ornl.gov/sites/publications/files/Pub55859.pdf`

[8] PyOrbit, `https://pyorbit-collaboration.github.io/`

[9] BMAD, `https://www.classe.cornell.edu/bmad/`

© Content from this work may be used under the terms of the CC BY 4.0 licence (© 2021). Any distribution of this work must maintain attribution to the author(s), title of the work, publisher, and DOI